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The Influence of External Load Configuration on Trunk Musculature and Spinal Stability  
during Manual Material Handling

Seyed Saman Madinei

Thesis submitted to the Benjamin M. Statler College of Engineering and Mineral Resources  
at West Virginia University in partial fulfillment of the degree of

Master of Science

in

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Configuration, Sudden Loading, Electromyography

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## **ABSTRACT**

### **The Influence of External Load Configuration on Trunk Musculature and Spinal Stability during Manual Material Handling**

**Seyed Saman Madinei**

The performance of manual material handling (MMH) tasks is highly associated with lower back injuries due to the excessive acute and/or cumulative mechanical loading that spinal tissues experience. Therefore, it is critical to understand how different characteristics of MMH tasks could become potential risk factors that change back injury risks and to develop proper MMH strategies that could reduce their biomechanical impacts to the spine. In this study, we explored the effects of external load configuration on trunk musculature and spinal stability during static loading and sudden loading scenarios.

The main objective of the current research was to explore how the configuration of an external weight (e.g. weight distribution or arrangement of the parts of the weight) can influence trunk biomechanics and spinal stability during the performance of static loading and sudden loading. To this end, we have conducted two experiments each of which was designed to simulate the two scenarios mentioned above.

In the first experiment, we investigated the influence of the weight configuration of hand loads on trunk muscle activities and the associated spinal stability during static weight holding. Thirteen volunteers each performed static weight holding tasks using two different 9 kg weight bars (with medial and lateral weight configurations) at two levels of height (low and high) and one fixed horizontal distance (result in constant spinal joint moment across conditions). Results of this study demonstrated that holding the laterally distributed load significantly reduced activation levels of

lumbar and abdominal muscles by 9 to 13% as compared with holding the medially distributed load.

In the second study, we examined the effects of different configurations of hand load on spine biomechanics and trunk stability during sudden loading events. Fifteen asymptomatic volunteers experienced sudden loadings using the same magnitude of weight (9 kg) with two different configurations (medially or laterally distributed) at three levels of height (low, middle, and high) and one fixed horizontal distance (constant spinal joint moment across conditions). Results of this study revealed that holding the medially distributed weight resulted in a significantly higher effective trunk stiffness (on average, lateral: 1785 Nm/rad and medial: 2413 Nm/rad) and peak L5/S1 joint compression force (on average, lateral: 2694 N and medial: 2861 N) compared with the laterally distributed weight.

We believe such effects are due to an elevated rotational moment of inertia when the weight of the load is laterally distributed. These findings suggest that during the design and assessment of manual material handling tasks such as lifting and carrying, the weight configuration of the hand load should be considered. According to the results, it was concluded that when confronted with static and sudden loading incidents, the load with larger moment of inertia (i.e. laterally distributed load) could help reduce the risk of low back injury compared to the load with a smaller moment of inertia (i.e. medially distributed load).

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## List of Nomenclatures

ANOVA	Analysis of Variance
CNS	Central Nervous System
EMG	Electromyography
EO	External Oblique
ES	Erector Spinae
LBP	Low Back Pain
MANOVA	Multivariate Analysis of Variance
MMH	Manual Material Handling
MU	Multifidus
MVC	Maximum Voluntary Contraction
NEMG	Normalized Electromyography
RA	Rectus Abdominus

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## CHAPTER 1. Introduction

### 1.1. Low Back Pain

The human spine is comprised of three major regions: cervical (7 vertebrae), thoracic (12 vertebrae), and lumbar (5 vertebrae). Below the lumbar region are the sacrum and coccyx vertebrae, which are fused together and form the sacrum and the coccyx [1]. Human spine can support the weight of the upper body, protect the spinal cord, maintain torso stability and allow for mobility and flexibility. There are three natural curves in the spine that give it an "S" shape when viewed from the side. These curves help the spine withstand great amounts of stress by providing a more even distribution of body weight. The morphology of the human spine determines that the lumbar region bears the bulk of the upper body's weight and transmits these loads to the pelvis and legs. As the lumbar spine provides greater mobility and flexibility compared to other spinal segments, it is also prone to greater stress. Therefore, all these characteristics have led to a clinically-noted increase in pain reporting and injuries associated with lumbar spine [2].

Low back pain (LBP) is the main and leading cause of disability worldwide [3] and is one of the most common reasons for missing work days [4, 5]. Epidemiological studies have demonstrated that approximately 80% of adult individuals will experience an episode of LBP at some point throughout their lives [6]. In the United States, the total costs of low back pain exceed \$100 billion per year; two-thirds of which are indirect, due to days-away-from-work and declined productivity [7]. Previous studies have demonstrated that some factors are associated with the onset of LBP. These factors are categorized into personal factors (e.g. genetics, smoking, overweight), psychosocial factors (e.g. emotional stresses, life and career dissatisfaction), and biomechanical factors (e.g. heavy lifting and lowering, frequent bending and twisting, sudden forceful incidents, prolonged static work postures) [8-12].

Findings of epidemiologic and biomechanical studies suggest that LBP is highly associated with heavy-duty work demand, awkward trunk postures, and frequent lifting, bending and twisting [13-16]. Work-related factors account for most of the LBP cases [17] and occupations exposing to overexertion, and bodily reaction (e.g. manual material handling) constitute the leading major event in the causation of LBP compared to other industries [18, 19]. Thus, understanding the risk factors which are associated with the performance of manual material handling (MMH) is of great importance in diminishing occupational LBP.

## 1.2. Stability of Spine

One of the major mechanical functions of the lumbar spine is to support the trunk by transmitting compression and shear forces to pelvis and legs during the performance of daily activities. It is well-documented that the isolated thoracolumbar spine would buckle under compressive forces exceeding 20 N and the lumbar section of the spine buckles under approximately 90 N of the compressive force [20, 21]. However, *in vivo*, the spinal column may experience compressive forces ranging from 6000 N for more demanding daily activities and up to 18000 N during competitive power lifting [22, 23]. It is clear that apart from the injuries associated with the lumbar spine exposure to excessive loads beyond tissue tolerance limits, the mechanical stability of the spinal column is of great concern at all levels of loading.

The stability of spine is achieved by a synergistic and coherent interplay of spinal subsystems including torso active and passive tissues, vertebrae, and central nervous system (CNS) [24-26]. It has been well-documented that the trunk musculature has a significant role to support the spine similar to guy wires spanning a bending mast [25, 27, 28]. Furthermore, the literature suggested

intra-abdominal pressure contributes to the overall stability of the spine [27-29]. There is also evidence of co-contraction of antagonistic torso muscles during daily activities, which is mechanically and energetically costly [30]. Interestingly, these co-contractions rise when a person prepares for an unexpected and sudden loading [31, 32]. Such co-contractions increase joint loads and muscular energy requirement for moment support. It has been postulated that a certain level of muscle co-contraction may be essential for the spine to maintain within a safe, stable, and equilibrium margin [24]. These stabilizing mechanisms are all controlled by the central nervous system. In order to prevent buckling of the spine, both the motor control system and the osteoligamentous spinal linkage will function within the range of mechanical stability, by holding some safety margin above the critical load [24]. Previous research maintained that the tasks which demand a high muscular effort prompt an ample spinal stability safety margin; while, lighter tasks render a declined stability safety margin to the spine [24, 33].

### 1.3. Risk factors associated with spinal stability

The previous research illustrated that in addition to excessive and prolonged compression and shear loading on the spinal vertebrae which could lead to LBP [12, 34-36], an unstable spine could also cause buckling and failure of spinal structures and result in spinal injuries [24, 37]. Any functional impairment of spinal tissues and unstable spinal motions (i.e. uncontrolled intervertebral movements), could lead to an improper distribution of mechanical loading to the spinal structure, causing buckling and failure of spinal structures, therefore elevates the risk of spinal injuries and pain [24, 38, 39].

Studies have found that the stability of spine and trunk muscular activity can be influenced by trunk posture [40, 41], lifting techniques [42, 43], work experience [44], prior knowledge of load magnitude [45] and prolonged working hours and the associated muscle fatigue [39, 46]. Granata and Wilson found out that the control of spinal stability was reduced in asymmetric postures associated with low-back disorder risk [40]. In another study it was propounded in comparison with the kyphotic postures, the lordotic postures increased the pelvic rotation, the active component of extensor muscle forces, segmental axial compression and shear forces at L5-S1, and spinal stability margin [41]. The findings of an effort also showed that squat lifting was advocated over stoop lifting as the technique of choice in reducing internal spinal loads [42]. Also, in order to achieve optimal carrying in terms of spine loading, one study suggested that loads should be positioned close to the body, even when carrying relatively light loads [43]. Plamondon et al. observed that the horizontal distance between the load and the person was smaller among expert workers compared to amateurs [44]. According to one study trunk muscle activity and sagittal acceleration were greater under unknown mass conditions which can be interpreted as reduced spinal stability [45]. The results of previous research also demonstrated that fatigue-related changes in muscle stiffness might reduce the capacity of the paraspinal muscles to stabilize the spine [39, 46]. It was also found that the magnitude of external loading and the asymmetry of the load handling posture (e.g. sagittally symmetric vs. asymmetric to the torso) could influence spinal stability and trunk muscle co-contraction [47, 48]. Also, handling load at different vertical heights could also influence spinal stability; previous studies have found reduced spinal stability when handling the load at a higher position (e.g. over shoulder level) as compared to a lower position (e.g. at waist level) [49, 50].



Sudden loading incident defined as an event in which the human body undergoes an abrupt force exertion is also considered as an influential factor in spinal stability, and it can be triggered by any destabilizing incidents (e.g. slips and falls) or impacts from external objects [51]. Furthermore, sudden loading is considered as a common incident in occupations with high exposure to unstable materials and body postures (e.g. construction, nursing, fishing, etc.) [52]. Previous literature has identified sudden loading as a major risk factor of LBP in the performance of MMH [53, 54]. The human body responds to sudden load release by naturally recruiting reflexive and voluntary muscle contractions to maintain the balance and whole body stability [32, 55]. Existing evidence reported an elevated activity level of both flexor and extensor torso muscles when the trunk encounters an unexpectedly applied flexion sudden load [56], most primarily to stiffen the trunk and augment spinal stability [38, 55, 57]. However, an increased co-contraction of back and abdominal muscles triggers higher compressive, and shear forces on the spine and could be a potential mechanism leading to LBP [32, 58]. Studies have hypothesized that either impaired motor control reactions by trunk musculature [26, 59, 60] or insufficient stabilization of lumbar spine before sudden loading exertion could trigger low back injuries [24, 26].

Previous literature has explored several contributing factors that could influence spinal stability and trunk biomechanics when the human body encounters sudden loading. Empirical measurements and theoretical analyses confirmed that during the sudden loading activity, fatigue development would alter the stiffness of muscles and would substantially increase trunk muscle activations in order to compensate for declined spinal stability [46]. One study found out that when experiencing sudden loading, lower handling positions would considerably reduce co-activation of torso muscles [50] which could be interpreted as augmented spinal stability [61, 62]. The influence of foot placement during unpredicted sudden loading on the biomechanics of spine has

also been investigated; Zhou and colleagues revealed that for the staggered standing posture, the activation of trunk muscles and the subsequent spinal loading declined significantly [63]. Another study reported an increased spinal loading and postural disturbances for an unpredicted and rapidly applied load in flexed trunk postures [64].

It was also noted that as the gap in the current knowledge, the configuration of an external load could potentially influence spinal stability during weight handling. Weight configuration (i.e. the distribution or arrangement of the parts of the weight) is a mechanical notion which can alter the moment of inertia of a hand load and could potentially influence spinal stability during weight handling. Changes in the weight distribution of a hand load result in changes in its moment of inertia; when performing manual material handling such change could cause the trunk to lose balance and lumbar spine to buckle especially under perturbation [65].

Figure 1 shows two different configurations of the same magnitude of weight on a weight bar. The rotational moment of inertia for each configuration can be calculated using Eq. (1), where  $I$  is the total moment of inertia for a specific weight around the axis of rotation.  $I_{CM_i}$  denotes the moment of inertia of component “ $i$ ” (e.g. a weight disc) around its center of mass;  $M_i$  represents the mass of the component “ $i$ ” and  $R_i$  is the distance between the center of mass of the component “ $i$ ” to the axis of the rotation. When discs are located further away from the midpoint of the bar, the system has larger rotational moment of inertia. As shown in Eq (2), where  $T$  is the total torque;  $I$  is the total moment of inertia and  $\alpha$  is the angular acceleration; for the same amount of torque (generated by external perturbation, unbalanced hand force etc.), a more laterally distributed load will experience smaller amount of angular acceleration and deviation, due to its larger moment of inertia as compared to a more medially distributed load. In mechanical systems, this concept is used to create more smooth motions; flywheels used in vehicle engines or industrial machines have

large moments of inertia to resist variations in applied torque and smooth out their mechanical output. Considering human biomechanics, this concept is also used to improve postural stability: when performing weight lifting weight discs are located on both sides of a barbell instead of at its center; during tightrope walking acrobats hold a long pole in order to enhance their whole body stability and balance [66].

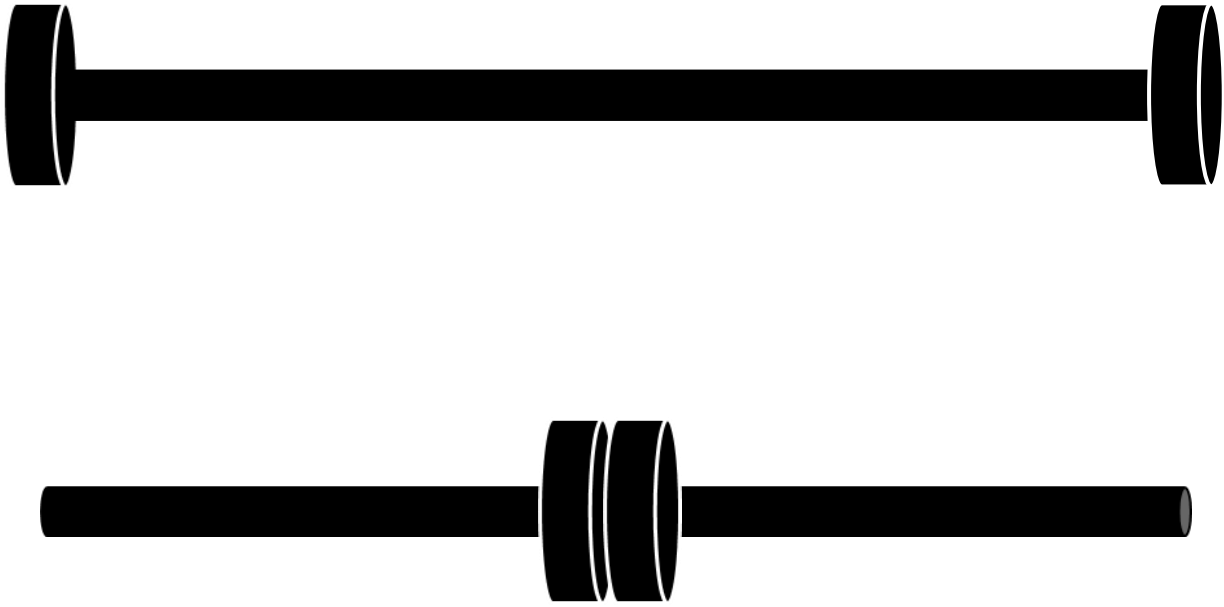


Figure 1. A demonstration of two objects with the same amount of weight but different configurations (the top panel shows a laterally distributed weight configuration, and the bottom panel shows a medially distributed weight configuration)

$$I = \sum_{i=1}^n (I_{CM_i} + M_i R_i^2) \quad (1)$$

$$T = I \times \alpha \quad (2)$$

## CHAPTER 2. Rationale and Objectives

A rigid mechanical structure, supporting two symmetric objects that have the same amount of weight but different configurations (as shown in Figure 1) should require the same and constant supporting forces at the points of contact. However, when these two weights are handled by humans, results can be quite different. The biomechanical system of the human body constantly relies on neuromuscular feedbacks to continuously adjust postures and force outputs even when performing quasi-static motions [67]. Previous studies declared that whether a task involves motions or not, the neuromuscular system of human body is always accompanied by a delay in understanding the signals representing an imbalance. As a result, corrective muscle reactions to this information are slowed down which may result in slight but prominent perturbations of body parts in quasi-static activities [66, 68]. When any disturbances occur either from the neuromuscular system (e.g. natural hand sway and slightly imbalanced hand force during static weight holding) or an external source (e.g. sudden load release), the weight with a greater moment of inertia will experience smaller angular acceleration and less movement [66].

The objective of this research is to investigate the influence of external load configuration on trunk musculature and spinal stability under the static loading and sudden loading scenarios. We will conduct two experiments to achieve the stated objectives. The first experiment focuses on the static performance of holding loads with different configurations, and it assesses the changes of trunk muscle activations and the associated trunk stability under these load configurations. The second experiment provides an assessment of spinal stability and trunk kinematics during the performance of sudden loading under the two different load configurations.

It was hypothesized that for the static loading scenario when handling the same magnitude of the external load, a more medially distributed load configuration would result in higher levels of

trunk muscle co-contraction as compared to a more laterally distributed load configuration. Furthermore, the hypothesis for sudden loading scenario was that under equal external moment, the weight with smaller moment of inertia (i.e. medially distributed weight) might undergo larger angular accelerations; This would diminish the spinal stability, thus higher trunk muscle contraction would be required to stabilize the load and the spinal loadings would be elevated consequently.

## CHAPTER 3. The Influence of External Load Configuration on Trunk Musculature and Spinal Stability during Static Weight Holding

### 3.1. Objective

The purpose of the current chapter is to understand the influence of the weight configuration of hand load on trunk muscle activities during static weight holding. Previous studies have found that during the performance of manual material handling tasks, to increase the stability of the spine, higher levels of trunk muscle co-contraction (i.e. an increase of both agonistic and antagonistic muscle activities) were often observed [69, 70]. Such mechanism could enhance the overall stiffness of the trunk and subsequently improve spinal stability [38, 61, 62, 71, 72]. Therefore, it was hypothesized that when handling the same magnitude of external loading, a more medially distributed load configuration would result in higher levels of trunk muscle co-contraction as compared to a more laterally distributed load configuration.

### 3.2. Method

#### 3.2.1. Participants

Thirteen healthy male volunteers (Average  $\pm$  SD) (age  $29.3 \pm 3.0$  years, body mass  $70.6 \pm 7.5$  kg, body height  $174.3 \pm 5.6$  cm) from the student population of West Virginia University participated in this study. Participants with a history of shoulder pain, low back pain or upper extremity injuries were excluded from this study. Prior to the data collection, informed consent forms were obtained from all participants. The experimental design and procedures were approved by the Institutional Review Board of West Virginia University.

### 3.2.2. Experimental Design

The weight configuration of the load and the load handling height were considered as two independent variables (referred to as CONFIGURATION and HEIGHT respectively from now on). CONFIGURATION has two levels: medially (9 kg weight located at the midpoint of the bar) and laterally distributed (two 4.5 kg weights located at the two ends of the bar with a distance of 100 cm). The load holding height also has two levels: high and low, which is defined as 25 cm above or below the shoulder height respectively for each participant. In order to maintain a constant external loading to the spine the distance between projected location of the center of mass of the load and the midpoint of participant's ankles remained at 45 cm throughout all trials [63]. Hand locations were also controlled; the mid-finger of each hand was 10 cm away from the mid-point of the bar for all trials. The combination of the two independent variables created four different conditions; each condition was repeated three times resulting in a total of 12 trials. Dependent variables included the EMG activities of Erector Spinae (ES), Multifidus (MU), Rectus Abdominus (RA) and External Oblique (EO) muscles.

### 3.2.3. Apparatus and equipment

A custom-made wooden stand was utilized as a guide to help participants control the height and the horizontal distance of the load during each trial. Weight discs were secured to two identical wood bars to create the two weight configuration conditions. Eight bipolar surface EMG electrodes (Bagnoli, Delsys, Inc., Boston, MA, USA) were placed over the skin of the left and right ES (4 cm lateral to the L3 spinous process), left and right MU (2cm lateral to the L5 spinous process) [73, 74] left and right RA (2 cm above and 3 cm lateral to the umbilicus), and left and right EO (approximately 15 cm lateral to the umbilicus) [75]. The sampling frequency of EMG was set at

1024 Hz. The placements of EMG electrodes are shown in Figure 2 (a) and (b). A lumbar dynamometer with a back flexion-extension module (Humac Norm, CSMi, MA, USA) was used in this study to secure participant's lower extremities and provide static resistance while performing maximum voluntary contractions (MVC) (described in more detail in the Procedure section).

#### 3.2.4. Procedure

Upon arrival, participants were given a thorough description of the experiment; then informed consent forms were signed. Prior to the data collection, participants performed a five-minute warm-up section to stretch shoulder and back muscles and become familiar with the protocol of the experiment. When finished, surface EMG electrodes were then attached to the above-described locations with double-sided tapes. During MVC trials, participants were secured in a 20° trunk forward flexion posture [76] and performed isometric trunk maximum voluntary flexion and extension contractions each with two repetitions. The 20° trunk forward flexion posture was selected because, in this posture, low back and abdominal muscles are close to their resting lengths which will allow them to generate maximal contractile tensions [77]. Each MVC trial lasted ~6 seconds and at least two minutes of rest was given between trials in order to avoid muscle fatigue [78, 79]. Participants then performed a total of 12 weight holding trials. Each trial required participants to hold a weight bar in front with a fixed horizontal distance, a predetermined vertical height (i.e. 25 cm above or below acromion points) and a fixed hand posture for 10 seconds (Figure 2 (c) and (d)). In each trial, weight was brought to and taken away from participants by an experimenter, therefore, no weight lifting, or lowering was needed for participants. At least one minute of rest was provided between trials in order to avoid shoulder and back muscle fatigue.



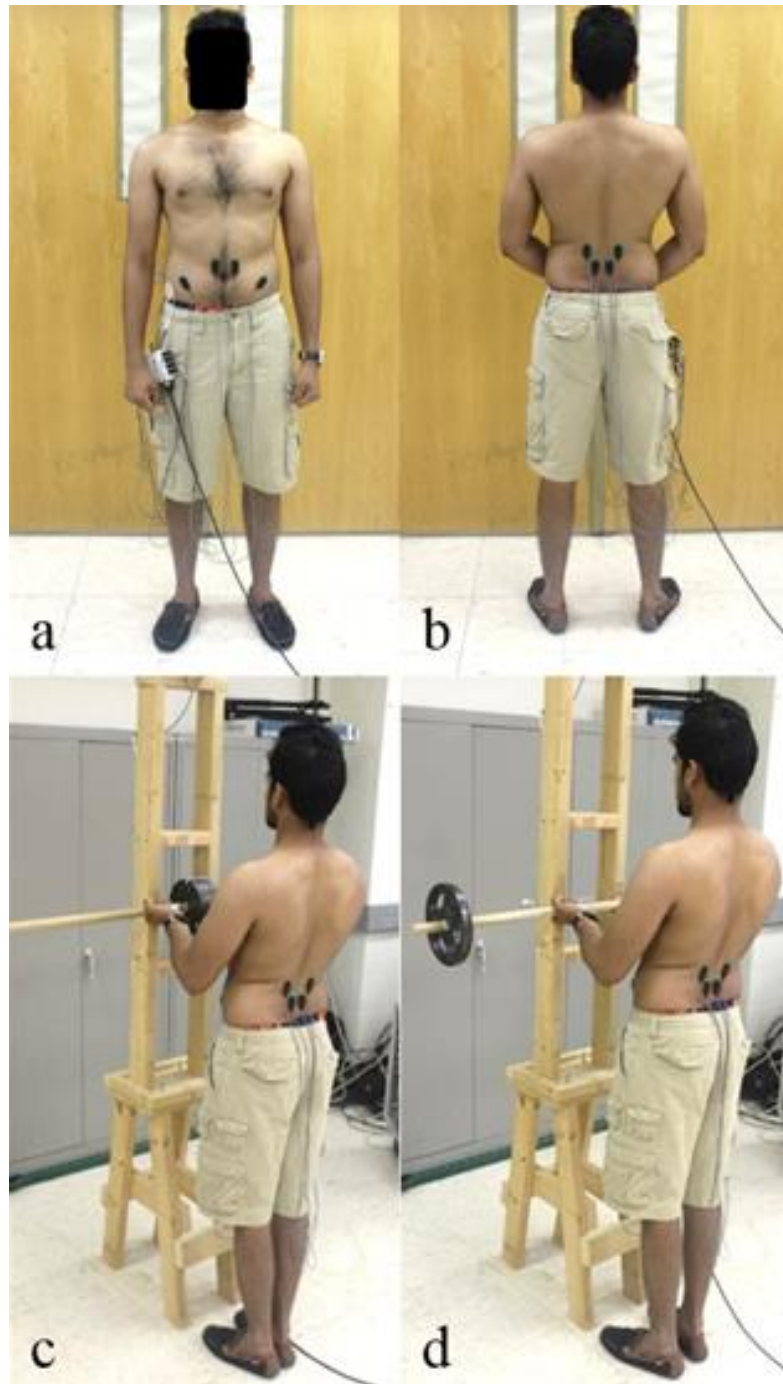


Figure 2. A demonstration of the location of EMG sensors (a, b), data collection apparatus and the posture participants used when holding a medially distributed weight (c) and a laterally distributed weight (d)

### 3.2.5. Data processing and analysis

EMG data were first filtered using a 10 Hz to 500 Hz band pass filter and a notch filter at 60 Hz and its aliases. The filtered EMG signals were then rectified and smoothed using a 100-data point sliding window. Mean EMG values from experimental trials were then normalized with respect to each muscle's maximal EMG collected from MVC trials. Due to the sagittally symmetric nature of the weight holding task, normalized EMG (NEMG) from left and right side of ES, MU, RA, and EO were averaged to generate NEMG for the correspondent muscle.

### 3.2.6. Statistical analysis

In total, each participant performed twelve randomized experimental runs (2 types of weight \* 2 height levels \* 3 repetitions) during the experiment. The three repetitions were considered separately and were not averaged together. A complete two-level factorial design with subjects considered as blocking was used in this experiment. The linear statistical model of this design is:

$$X_{ijkl} = \mu + \tau_i + \beta_j + (\tau\beta)_{ij} + \delta_k + \varepsilon_{ijkl} \quad (3)$$

Where,

$\mu$  denoted the overall mean of all observations.

$\tau_i$  denoted the effect of weight configuration with two levels ( $i = 1$  and  $2$ ).

$\beta_j$  denoted the effect of load handling height with two levels ( $j = 1$  and  $2$ ).

$(\tau\beta)_{ij}$  denoted the interaction effect of the weight configuration and load handling height.

$\delta_k$  denoted the effect of the block, represents the number of participants ( $k = 1, \dots, 13$ ).

$\varepsilon_{ijkl}$  was a random error term.

$X_{ijkl}$  represented the NEMG of each pair of trunk muscles for each replication ( $l=1, 2, 3$ ).

The weight configuration ( $\tau_i$ ) and height ( $\beta_j$ ) were treated as fixed variables and subjects ( $\delta_k$ ) were treated as a random block. The random error term,  $\varepsilon_{ijk}$  was assumed to follow normally and independently distributed (NID) ( $0, \sigma^2$ ) in the model. The four dependent variables were the NEMG of each pair of trunk muscles (ES, MU, EO, and RA). The appropriate F tests were applied to test the model significance and the individual effect of the factors and their interactions. The hypotheses of interests were:

$$H_{10}: \tau_i = 0;$$

$$H_{1A}: \text{at least one } \tau_i \neq 0.$$

$$H_{20}: \beta_j = 0;$$

$$H_{2A}: \text{at least one } \beta_j \neq 0.$$

The following power approach was implemented in order to determine the appropriate sample size:

$$\text{Test for interaction: } \varphi = \frac{1}{\sigma} \sqrt{\frac{n \sum_{i=1}^2 \sum_{j=1}^2 (\tau\beta)_{ij}^2}{(a-1)(b-1)+1}} \quad (4)$$

$$v_1 = (a-1)(b-1), v_2 = ab(n-1)$$

$$\text{Test for weight configuration: } \varphi = \frac{1}{\sigma} \sqrt{\frac{nb \sum_{i=1}^2 \tau_i^2}{a}} \quad (5)$$

$$v_1 = a-1, v_2 = ab(n-1)$$

$$\text{Test for load handling height: } \varphi = \frac{1}{\sigma} \sqrt{\frac{na \sum_{i=1}^2 \beta_j^2}{b}} \quad (6)$$

$$v_1 = b - 1, v_2 = ab(n - 1)$$

Where,  $a=2$  and  $b=2$ .

The operating characteristic (OC) curves were used to assist in iterating the number of replications (n) to achieve the desired power value (0.9).

All statistical analyses were performed using Minitab 17 (Minitab Inc., PA, USA). The validity of ANOVA assumptions (normality of residuals, equality of variances, etc.) was tested on all dependent variables and the ones which did not satisfy those assumptions were transformed until all assumptions were satisfied [80]. Multivariate ANOVA (MANOVA) analysis was subsequently performed to assess the effects of main effects and their interactions on all dependent variables collectively. Significant effects were further tested using univariate ANOVA. The level of significance,  $\alpha = 0.05$  was chosen for all statistical analyses.

### 3.3. Results

In agreement with our initial hypothesis, the results of MANOVA revealed significant main effect of both independent variables: CONFIGURATION and HEIGHT. The interaction effect was not significant and therefore was not further analyzed (Table 1). Results of univariate ANOVA illustrated that both independent variables substantially influenced all back and abdominal muscles. As shown in Figure 3 and 4, higher NEMG values were observed in higher load handling position and with medial load condition for all trunk muscles. Whereas lower load handling position and laterally distributed load generated lower NEMG values. More specifically, when holding a medially distributed load, NEMG of ES and MU muscles went up 2.0% (24.0% to

26.0%) and 3.0% (29.7% to 32.7%) respectively, abdominal muscle activity also went up 0.5% (5.4% to 5.9%) and 1.0% (8.6% to 9.6%) respectively for RA and EO. Holding the load at a higher position resulted in 3.8% (23.0% to 26.8%), 3.5% (29.4% to 32.9%), 1.5% (5.0% to 6.5%) and 1.5% (8.3% to 9.8%) increase in NEMG for ES, MU, RA and EO muscles respectively as compared with holding the load at a lower position.

Table 1. The results of MANOVA and Univariate ANOVA. Bold values indicate significant p-values.

Independent variables	MANOVA	ANOVA			
		ES	MU	EO	RA
Height	<b>&lt;0.001</b>	<b>&lt;0.001</b>	<b>&lt;0.001</b>	<b>&lt;0.001</b>	<b>&lt;0.001</b>
Configuration	<b>&lt;0.001</b>	<b>&lt;0.001</b>	<b>&lt;0.001</b>	<b>&lt;0.001</b>	<b>0.012</b>
Height × Configuration	0.086	N/A	N/A	N/A	N/A

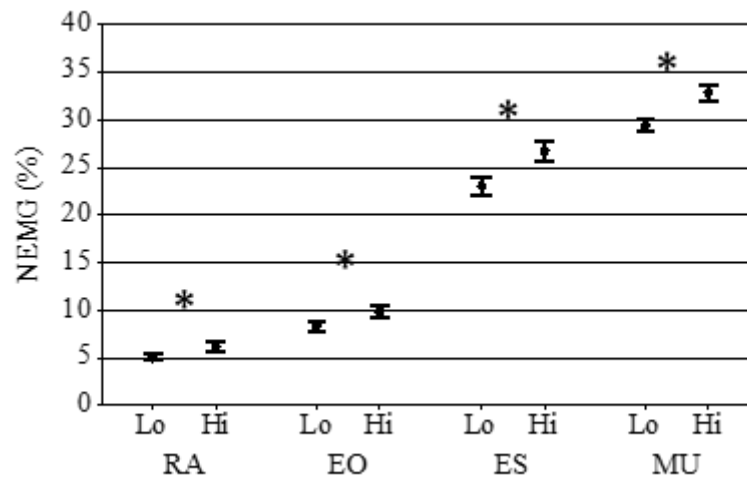


Figure 3. Averaged activation levels of trunk muscles when holding the load in high (Hi) vs. low (Lo) conditions. Bars indicate the corresponding standard error and asterisks denote statistical significance between two levels

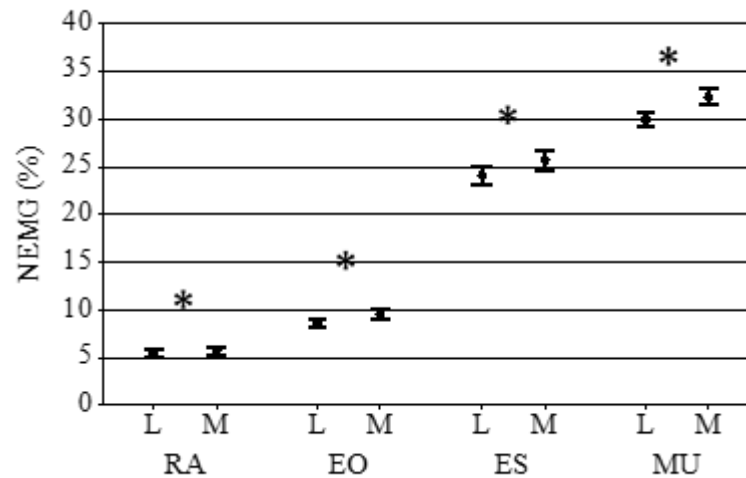


Figure 4. Averaged activation levels of trunk muscles when holding a laterally distributed weight (L) vs. a medially distributed weight (M). Bars indicate the corresponding standard error and asterisks denote statistical significance between two levels

### 3.4. Discussion.

The main purpose of current study was to understand whether different weight configurations of the hand load could influence trunk muscle activities and the associated spinal stability during weight holding. As described earlier, when holding the same magnitude of the load, different weight configuration could generate a different rotational moment of inertia. Such difference could result in changes in neuromuscular control and muscle activation patterns in order to cope with external perturbations or unbalanced hand forces.

We hypothesized that when handling the same magnitude of external loading, a more medially distributed load configuration would result in higher levels of trunk muscle co-contraction as compared to a more laterally distributed load configuration. Results of the current study

demonstrated increased EMG activities among lumbar extensor muscles as well as abdominal muscles; these results supported our initial hypotheses. Previous research showed that signals representing imbalance are generated and understood by the brain with a delay, and corrective muscle reactions to this information are consequently slowed down which may result in slight perturbations of body parts [66, 68]. It can be understood that because of the existence of neuromuscular delay (i.e. difference in time between the onsite of hand load change and the response in change of hand force), hand load that had smaller moment of inertia (i.e. medially distributed load) could generate greater angular acceleration and displacement to the load, in turn, larger trunk muscle forces would be needed to stabilize the system and bring the load back to its equilibrium.

Figure 5 demonstrates two typical EMG profiles of an ES muscle during two weight holding trials. As can be observed, when holding a laterally distributed load we often recorded much more stable EMG activities as compared to holding a medially distributed load. Also, in agreement with previous studies [49, 50], higher levels of trunk muscle activities were observed when holding a load in a higher position. Such increase of trunk muscle co-contraction is possibly used to increase the stiffness of the trunk and compensate for the reduced trunk and spinal stability [38, 49]. Another interesting finding of this study was that in the abdominal region, EO muscles demonstrated higher NEMG in comparison to RA muscles. We believe this is mainly due to the differences in the muscle fiber orientation and the location of these two muscles. RA is located in the center of the abdomen, and its muscle fibers are vertically oriented whereas EO is located laterally to the RA and its muscle fibers are oriented at a slightly oblique angle [81]. This location and muscle fiber orientation enabled EO muscle to play an important role in balancing rotational moment such as what is needed in the current experiment.

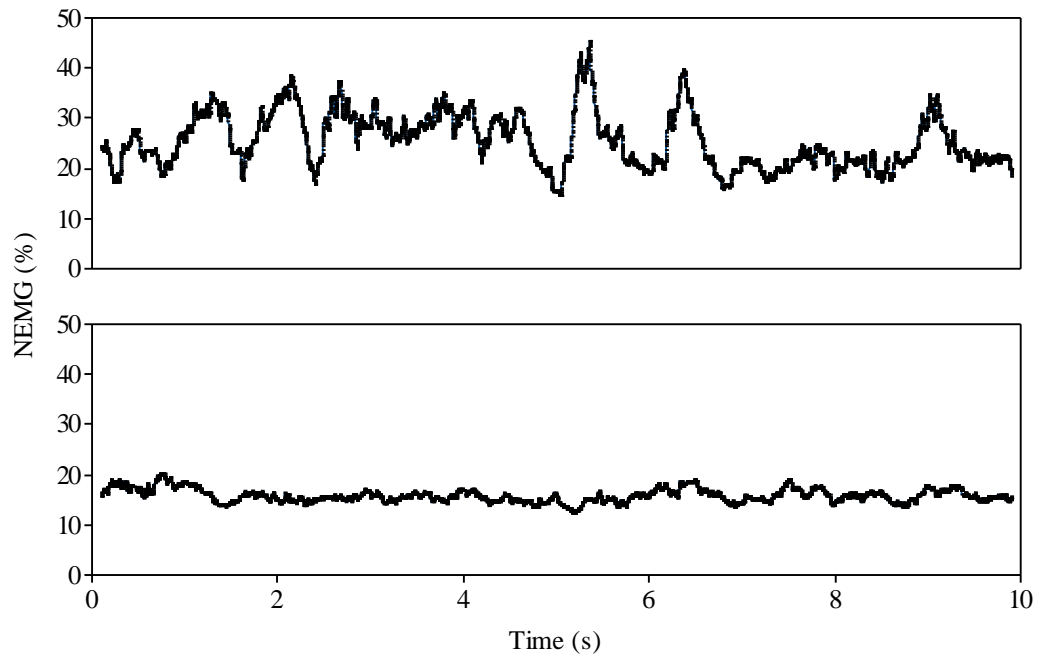


Figure 5. A demonstration of typical EMG profiles of an ES muscle when holding a medially distributed weight (top panel) vs. a laterally distributed weight (bottom panel)

Although both main effects significantly influenced trunk muscle activities during weight holding, the actual differences between levels were relatively small; for example, compared to the medially distributed load configuration, holding a laterally distributed load reduced the NEMG of ES, MU, EO and RA muscles by 9%, 10%, 13% and 9% respectively, however, the actual changes in magnitude were only 2.0%, 3.0%, 1.0% and 0.5% of the maximal EMG values respectively. We believe this observed small differences between levels were mainly because our testing conditions were restricted to relatively simple, conservative and controlled conditions. For instance, to avoid back and shoulder muscle fatigue, the horizontal distance of the load was controlled at 45 cm across all trials, and the duration of each weight holding task was limited to only 10 seconds. Also, to further reduce variance, we used static weight holding task instead of



dynamic motions. It is highly possible that in real working scenarios where workers are required to perform dynamic tasks, with more demanding task requirements and the possibility of experiencing muscle fatigue and external perturbations, the influence of weight configuration on trunk muscle activation levels and spinal stability can be much magnified.

Finally, some limitations of this study need to be noted: first, task duration was controlled at 10 seconds in order to avoid shoulders and trunk muscle fatigue; longer task duration and the associated muscle fatigue could further influence the results which warrant future investigation. Second, only one level of weight and one specific hand and trunk posture were tested in the current study, heavier weight and more complex hand and body postures (e.g. asymmetric weight holding) were not explored. Third, only young, male participants were recruited. Older individuals and female participants may generate different results.

## CHAPTER 4. The Influence of External Load Configuration on Trunk Musculature and Spinal Stability during Unexpected and Sudden Load Release

### 4.1. Objective

The objective of this experiment was to examine the effects of different configurations of hand load on spine biomechanics and trunk stability during unexpected sudden loading events. Based on the results of previous studies it was hypothesized that during the performance of sudden loading and under the same amount of external moment, the weight with a smaller moment of inertia (i.e. medially distributed weight) might undergo larger angular accelerations and displacement, therefore increase the mechanical loading of the spine and reduce spinal stability.

## 4.2. Method

### 4.2.1. Participants

Fifteen asymptomatic male volunteers from student population of West Virginia University with mean (SD) age, mass and height of 26.3 (4.9) years, 74.6 (8.5) kg, and 176.5 (5.2) cm respectively, participated in this study. Participants reported no history of shoulder and low back pain, upper/lower limb injuries and they had no preceding training or working experience of manual material handling. An informed consent form was obtained from each volunteer before the data collection session. The experimental procedure and design for this study were approved by the Institutional Review Board of West Virginia University.

### 4.2.2. Experimental Design

The load handling height (referred to as HEIGHT) and the weight configuration of the load (referred to as CONFIGURATION), were considered as independent variables in this experiment. The HEIGHT consisted of three levels: low (25 cm below the shoulder height) middle (at the shoulder height) and high (25 cm above the shoulder height). The CONFIGURATION included two levels: medial (9 kg weight disc located at the midpoint of the barbell) and lateral (two 4.5 kg weight discs each located at the sides of the barbell with 100 cm distance) (Figure 1). The combination of two CONFIGURATION and three levels of HEIGHT constituted six different conditions and each participant performed four repetitions for each of the six conditions resulting in a total of 24 trials.

Three dependent variables which were investigated in this study included: 1. The increase of trunk flexion angle: the maximum increase of trunk flexion angle during the sudden loading event

relative to the initial trunk posture; 2. Peak L5/S1 joint compression force: the maximum compressive force on the L5/S1 intervertebral disc during the sudden loading event. 3. Lumbar stiffness coefficient (K) determined from a second-order biomechanical model described in section 4.2.7.2.

#### 4.2.3. Apparatus and equipment

A custom-made wood structure was built which served as a guide for participants to maintain the horizontal distance of the load and adjust their performances according to the specified heights. Weight discs were tightly secured to two identical wooden bars and constituted the two configuration conditions with equal weights of 9 kg (i.e. medial and lateral). Eight bipolar surface electromyography (EMG) electrodes (Bagnoli, Delsys, Inc., Boston, MA, USA) were utilized to record the myoelectric activity of 8 trunk muscles at a sampling frequency of 1024 Hz. These electrodes were placed over the skin of bilateral ES (4 cm lateral to the L3 spinous process) [79], bilateral MU (2cm lateral to the L5 spinous process) [73, 74], bilateral EO (10 cm lateral to the umbilicus and 4 cm above the ilium with the angle of 45°), and bilateral RA (2 cm above and 3 cm lateral to the umbilicus) [79]. The locations of electrodes are presented in Figure 6 (a) and (b). Trunk and upper extremities kinematics were captured by an eight-camera (MX-13 series) 3D optical motion tracking system (Vicon, Nexus, Oxford, UK) at a sampling frequency of 100 Hz. Eleven reflective markers were placed over the spinous processes of C7, T12, and L5 vertebrae, the most dorsal point on acromioclavicular joint of the left and right shoulders, the most caudal point of the lateral epicondyle of the left and right elbows, the ulnar side of both wrists and the distal side of the third metacarpal bone on both hands [50]. The EMG and kinematics data were recorded and synchronized by Nexus 10.7 software (Vicon, Nexus, Oxford, UK). A lumbar dynamometer (Humac Norm, CSMi, MA, USA) with a back flexion module was used in order to

secure participants' pelvis and lower extremities and also provide a static resistance during the trunk maximum voluntary contraction (MVC) trials.

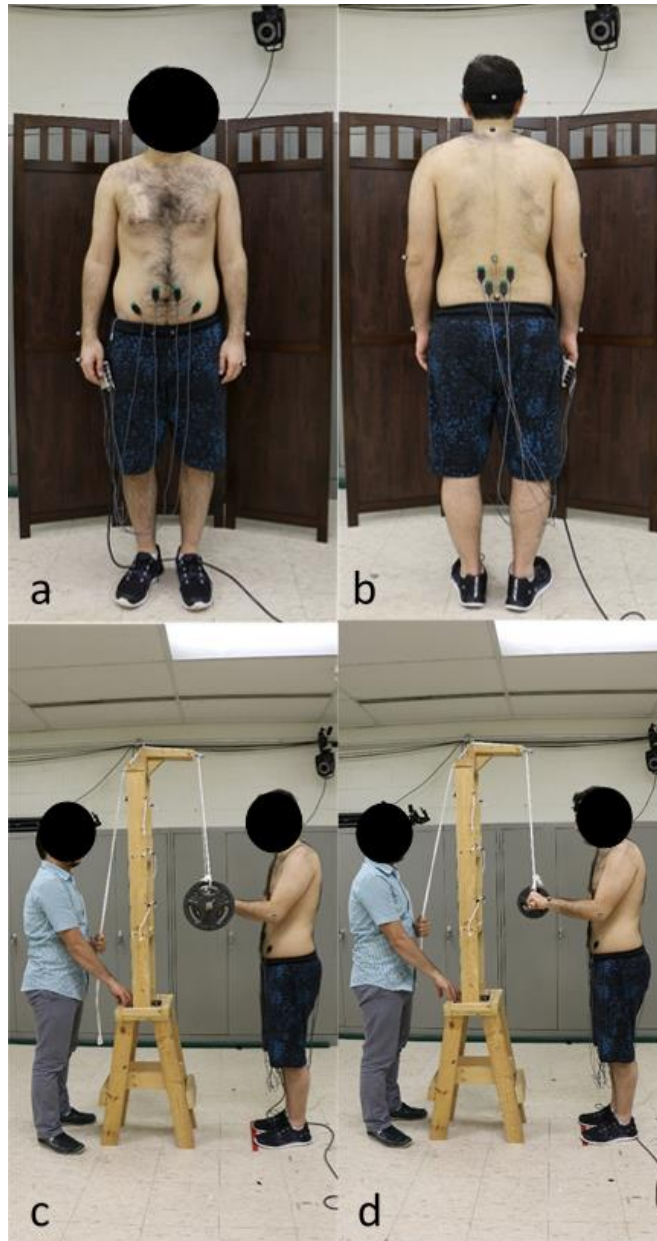


Figure 6. A demonstration of the location of EMG sensors (a, b), side view of the data collection apparatus and the postures participants used when undergoing the sudden load release of a laterally distributed weight at low height (c), a medially distributed weight at low height (d)

#### 4.2.4. Procedure

A complete description of the experimental protocol was provided for participants upon their arrival, and informed consent forms were then signed and obtained. After the measurement of participants' anthropometric data including body height, weight, trunk length (the vertical distance from L5/S1 joint to the top of the head), trunk width (at iliac and xiphoid process levels) and trunk depth (at iliac and xiphoid process levels), a brief warm-up session was provided for them to stretch out their trunk and shoulder muscles and get familiarized with the experiment protocol. Eight bipolar surface EMG electrodes were then attached to the skin of the muscles mentioned above (swabbed by alcohol) with double-sided tapes. At the beginning of the experiment, participants were secured in a 20° trunk forward flexion posture and performed two repetitions of isometric maximum trunk flexion/extension exertions against a static resistance provided by the dynamometer [76]. Each MVC trial lasted for 6 sec and 2 minutes of rest was provided between the exertions to avoid muscle fatigue [78, 79].

Following that, eleven reflective markers were attached to the above-described locations of the trunk upper limbs with double-sided tapes. Volunteers at the beginning performed three repetitions of flexion/extension trials consisting of flexing for 6 sec, maintaining in full flexion posture for 5 sec and extending for 6 sec. These data was later on used to obtain the gain value for each subject. Afterward, they performed a total of 24 trials in a completely randomized order. Each trial started with subjects closing their eyes, standing with feet shoulder width apart and gripping the barbell without supporting its weight; the load was then suddenly released by the experimenter with no preceding notice (Figure 6 (c) and (d)). Subjects were then required to open their eyes, immediately grasp the falling load, bring it back to its initial position, and stably hold it for 3 seconds. The subject then handed back the loads to the experimenter, and he put them on the

ground. There was at least one minute of rest provided between the trials to prevent shoulder and back muscle fatigue. In order to maintain the constant external flexor moment applied to the spine, participants were instructed to stand straight in front of the wooden structure such that the horizontal distance between the midpoint of their ankles and the projected location of the center of mass of the load remained at 45 cm throughout the whole tasks [33]. Throughout the experiment, the locations of hand grip remained consistent such that there was a 20 cm distance between the third metacarpals of both hands and arms remained in prone condition.

#### 4.2.5. Data processing and analysis

EMG signals were primarily filtered using a band pass filter between 10 Hz to 500 Hz and a notch filter at 60 Hz and its aliases. The filtered signals were then rectified and smoothed with a 100-data point (~0.1 sec) sliding window. The EMG data recorded from all experimental trials were subsequently normalized to the EMG activities obtained during the MVC trials for each muscle. Following that, trunk and upper extremity kinematics were calculated using extracted 3D coordination data from the eleven reflective markers. Trunk flexion angle was calculated as the angle between trunk segment (the line connecting C7 and L5 reflective markers in the sagittal plane) and the transverse plane [73].

#### 4.2.6. Statistical analysis

In total, each participant performed twenty four randomized experimental runs (2 types of weight \* 3 height levels \* 4 repetitions) during the experiment. The four repetitions were

considered separately and were not averaged together. A general full factorial design with subjects considered as blocking was used in this experiment. The linear statistical model of this design is:

$$X_{ijkl} = \mu + \tau_i + \beta_j + (\tau\beta)_{ij} + \delta_k + \varepsilon_{ijkl} \quad (7)$$

Where,

$\mu$  denoted the overall mean of all observations.

$\tau_i$  denoted the effect of weight configuration with two levels ( $i = 1$  and  $2$ ).

$\beta_j$  denoted the effect of load handling height with two levels ( $j = 1, 2$ , and  $3$ ).

$(\tau\beta)_{ij}$  denoted the interaction effect of the weight configuration and load handling height.

$\delta_k$  denoted the effect of the block, represents the number of participants ( $k = 1, \dots, 15$ ).

$\varepsilon_{ijkl}$  was a random error term.

$X_{ijkl}$  represented the peak L5/S1 joint compression force, trunk stiffness and maximum trunk flexion angle individually for each replication ( $l = 1, \dots, 4$ ).

The weight configuration ( $\tau_i$ ) and height ( $\beta_j$ ) were treated as fixed variables and subjects ( $\delta_k$ ) were treated as a random block. The random error term,  $\varepsilon_{ijk}$  was assumed to follow normally and independently distributed (NID) ( $0, \sigma^2$ ) in the model. The three dependent variables were the peak L5/S1 joint compression force, trunk stiffness and maximum trunk flexion angle. The appropriate F tests were applied to test the model significance and the individual effect of the factors and their interactions. The hypotheses of interests were:

$$H_{10}: \tau_i = 0;$$

H<sub>1A</sub>: at least one  $\tau_i \neq 0$ .

H<sub>20</sub>:  $\beta_j = 0$ ;

H<sub>2A</sub>: at least one  $\beta_j \neq 0$ .

The following power approach was implemented in order to determine the appropriate sample size:

$$\text{Test for interaction: } \varphi = \frac{1}{\sigma} \sqrt{\frac{n \sum_{i=1}^2 \sum_{j=1}^2 (\tau\beta)_{ij}^2}{(a-1)(b-1)+1}} \quad (8)$$

$$v_1 = (a-1)(b-1), v_2 = ab(n-1)$$

$$\text{Test for weight configuration: } \varphi = \frac{1}{\sigma} \sqrt{\frac{nb \sum_{i=1}^2 \tau_i^2}{a}} \quad (9)$$

$$v_1 = a-1, v_2 = ab(n-1)$$

$$\text{Test for load handling height: } \varphi = \frac{1}{\sigma} \sqrt{\frac{na \sum_{i=1}^2 \beta_j^2}{b}} \quad (10)$$

$$v_1 = b-1, v_2 = ab(n-1)$$

Where,  $a=2$  and  $b=3$ .

The operating characteristic (OC) curves were used to assist in iterating the number of replications (n) to achieve the desired power value (0.9).

All statistical analyses of the current study were performed by Minitab 17 software (Minitab Inc., PA, USA). The ANOVA assumptions were primarily validated (e.g. normality of residuals, equality of variances, etc.) and the dependent variables which violated the assumptions were transformed using deterministic mathematical functions until all assumptions were satisfied [80].



Multivariate analyses of variance (MANOVA) was subsequently conducted to reveal the statistical significance of main effects, CONFIGURATION, and HEIGHT, as well as their interactions on all dependent variables collectively. Significant effects were further analyzed using repeated measures univariate ANOVA with ‘subject’ considered as a blocking factor. Tukey-Kramer post-hoc tests were conducted on dependent variables that were significantly affected by HEIGHT to further investigate the differences between each two HEIGHT levels. The level of significance,  $\alpha = 0.05$  was chosen for all statistical analyses.

#### 4.2.7.1. EMG-assisted Biomechanical Model

To estimate the external moment exerted on the L5/S1 intervertebral joint of the spine, a multi-segment dynamic model consisting of seven body segments (trunk, left and right upper arms, forearms, and hands) and the external load was recruited. Masses and centers of mass of body segments were estimated similar to the methods recommended in the existing literature [82]. Muscle forces and the corresponding internal moment and spinal compression force at the L5/S1 joint were calculated using an established and validated EMG-assisted biomechanical model [83]. This model estimated the internal moment at the L5/S1 joint using Eq. 11, where “ $Gain$ ”, “ $NEMG_i$ ” and “ $A_i$ ” respectively denote muscle gain value, the normalized EMG with respect to the maximum voluntary contraction (MVC), and the cross-sectional area of muscle  $i$ ; “ $f(l_i)$ ” and “ $f(v_i)$ ” are the muscle force-length and force-velocity modulation factors of muscle  $i$  respectively and “ $r_i$ ” represents the moment arm vector of muscle  $i$  [83, 84]. The joint compression force at the L5/S1 level of the spine was calculated based on trunk muscle forces, geometry (e.g. line of action) and kinematics. The anthropometric data (mass, height, trunk length, width, and depth) obtained from each participant helped to estimate the subject-specific moment arms and cross-sectional areas of

the trunk muscles incorporated in the model, using predictive equations from the literature [85, 86]. The gain value (maximum muscle stress) was then derived from matching the internal and external moments at the static period of load holding (last 3 sec) of all trials.

$$\vec{M} = \sum_{i=1}^8 \vec{r}_i \times \text{Gain} \times NEMG_i \times A_i \times f(l_i) \times f(v_i) \quad (11)$$

#### 4.2.7.2. Trunk Stiffness Biomechanical Model

Trunk stiffness can be calculated using the kinematics data of trunk response to a sudden load release (quick release protocol) [87-91]. In a model developed by Cholewicki and colleagues, the trunk was represented as a second-order system with viscoelastic properties, oscillating freely around the L5/S1 joint after a sudden load release [71] (Figure 7). The frequency and amplitude of such oscillations recorded promptly after the load release, but before the occurrence of voluntary muscle contraction, are determined by the trunk inertia ( $I$ ), damping coefficient ( $B$ ) and stiffness coefficient ( $K$ ) established before the load release. For small trunk angles ( $\theta$ ):

$$I\ddot{\theta} + B\dot{\theta} + K(\theta - \theta_0) = mgL \sin \theta, \quad (12)$$

where  $mg$  is trunk and external load,  $L$  is the vertical distance from the center of trunk mass supposed to be at T9 level to the L5/S1 joint, and  $\theta_0$  is a hypothetical resting angle of the rotational spring. Subject-specific trunk mass (including head and arms) and moment of inertia were calculated from the anthropometric data [92]. The mass of the external load has been included in calculation of total trunk inertia [71]. A curve-fitting algorithm was implemented to acquire the best match between the measured and modeled trunk deflection trajectories and consequently determine coefficients  $B$ ,  $K$  and a constant  $C$  (encompassing  $\theta_0$  and integration constants). This

protocol was applied to a double integration of Eq. (12), since according to previous research integration is numerically a more robust operation than differentiation [90], thus:

$$I\ddot{\theta} + B \int \dot{\theta} dt + K \iint \theta dt^2 + Ct^2 = mgL \iint \sin \theta dt^2 \quad (13)$$

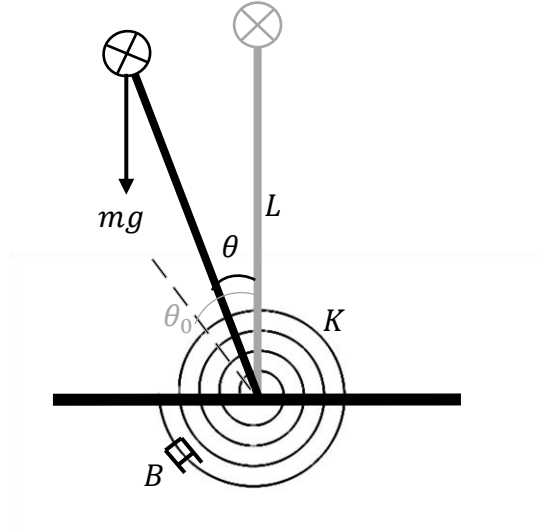


Figure 7. A free body diagram of a second-order trunk model oscillating around the L5/S1 joint after a sudden load release

It has been shown that the minimum data length to accurately identify parameters in Eq. (13) is equivalent to at least a quarter of wavelength [71]. Thus, trunk flexion angle data, taken from the time of load release to the point of maximum trunk deflection (average = 441 ms, S.D. = 97 ms) was used for the curve fitting procedure (Fig. 8A and 8B). This time interval was sufficiently short to exclude voluntary muscle contractions [93]. However, involuntary muscle reflex responses can take place between 40-80 ms following the load release [60]. Thus, the quantification of trunk stiffness obtained from the expressed method was an effective stiffness, combining pre-set trunk stiffness and reflex response.

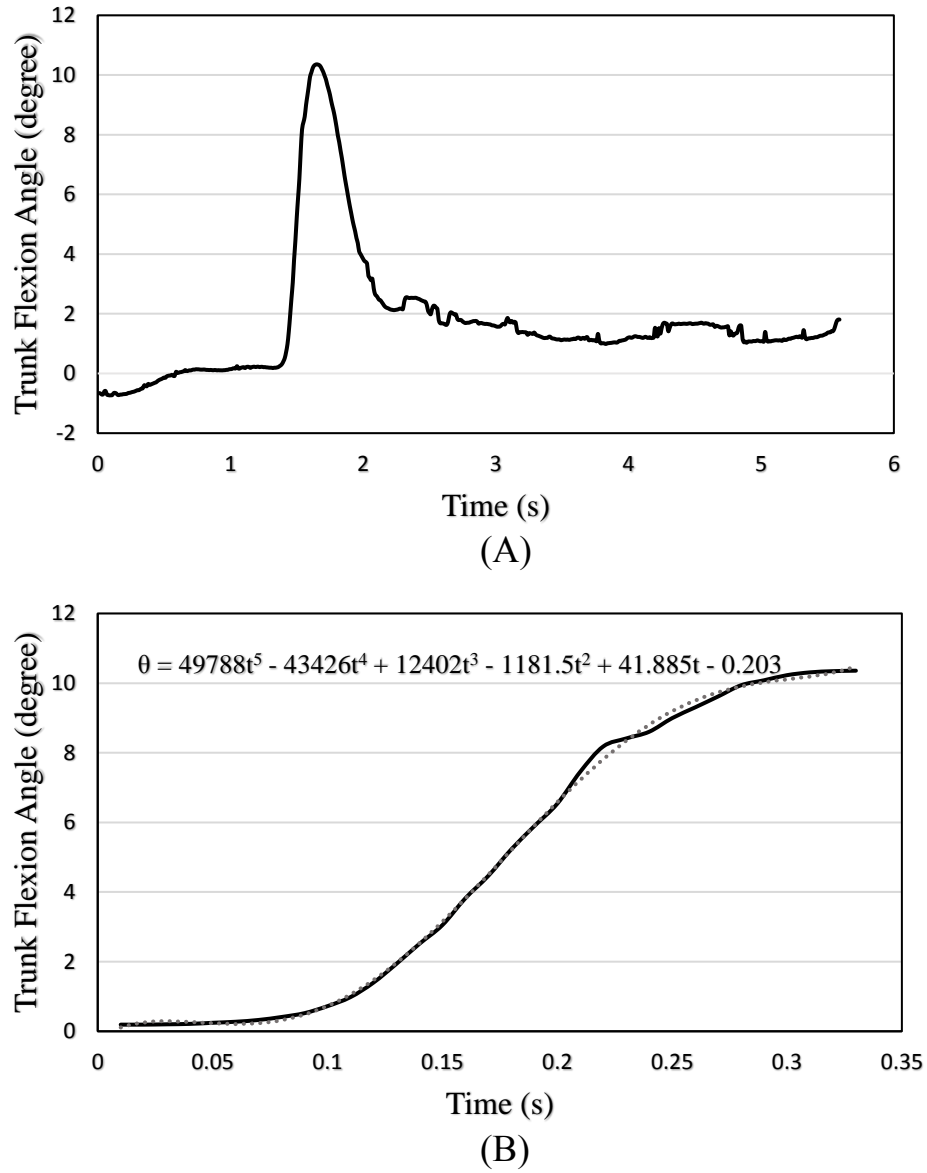


Figure 8. A. An example of trunk motion response to a sudden load release. B. Fitting the best polynomial to the quarter of the wavelength (solid line represents experimental data, and dot line represents the best-fitted curve). Data belong to one participant performing a trial at low height and laterally distributed condition

### 4.3. Results

The results of MANOVA revealed significant main effects of both independent variables (i.e. CONFIGURATION and HEIGHT) on all dependent variables (i.e. peak compression force, trunk stiffness, and maximum trunk flexion angle). However, their interaction effect was not significant and thus was not further analyzed. The results of univariate ANOVA demonstrated that HEIGHT considerably affected all dependent variables. According to these results, the peak compression force and trunk stiffness were significantly influenced by CONFIGURATION.

The effects of HEIGHT on all dependent variables are depicted in Figs. 9-11. A significant rise of peak L5/S1 joint compression force was observed with an increase of load handling height (on average from 2453 N to 3175 N) (Fig. 9). Effective trunk stiffness considerably elevated with an increase of load handling height (on average from 1552 Nm/rad to 2538 Nm/rad) (Fig. 10). Finally, a significantly smaller maximum trunk flexion angle ( $11.3^{\circ}$ ) was observed at the “High” handling position, while the difference between “Low” and “Middle” conditions was not statistically significant (Fig. 11).

The effects of CONFIGURATION on all dependent variables are also shown in Figs. 12-14. Holding the medially distributed weight resulted in a higher peak compression force compared to the laterally distributed weight (on average, lateral: 2694 N and medial: 2861 N) (Fig. 12). Higher effective trunk stiffness was observed in medially distributed weight condition in comparison to laterally distributed weight (on average, lateral: 1785 Nm/rad and medial: 2413 Nm/rad) (Fig 13). Maximum trunk flexion angle was not considerably affected by CONFIGURATION (on average, lateral:  $12.2^{\circ}$  and medial:  $12.6^{\circ}$ ) (Fig. 14).

Table 2. The results of MANOVA and univariate ANOVA. Bold values indicate significant p-values.

Independent Variables	MANOVA	ANOVA		
		Compression Force	Trunk Stiffness	Max Trunk Flexion
Configuration	<b>&lt;0.001</b>	<b>&lt;0.001</b>	<b>&lt;0.001</b>	0.191
Height	<b>&lt;0.001</b>	<b>&lt;0.001</b>	<b>&lt;0.001</b>	<b>&lt;0.001</b>
Configuration*Height	0.479	N/A	N/A	N/A

Table 3. The mean (SD) values of dependent variables at different HEIGHT and CONFIGURATION conditions.

Dependent Variables	Low		Middle		High	
	Lateral	Medial	Lateral	Medial	Lateral	Medial
Peak L5/S1 Compression Force (N)	2348 (596)	2559 (598)	2640 (585)	2767 (593)	3095 (698)	3256 (613)
Trunk Stiffness (Nm/rad)	1305 (900)	1799 (1073)	1918 (1089)	2496 (1069)	2133 (1055)	2943 (1232)
Maximum Trunk Flexion Angle (degree)	12.3 (4.3)	13.0 (3.6)	13.1 (4.0)	13.5 (4.4)	11.3 (4.4)	11.4 (4.6)

Figures 9-11 respectively depict peak L5/S1 compression force, trunk stiffness, and maximum trunk flexion angle caused by sudden loading under the three different load handling heights (two CONFIGURATION levels combined). Different letters denote levels that are statistically different from one another. Bars indicate the corresponding standard error.

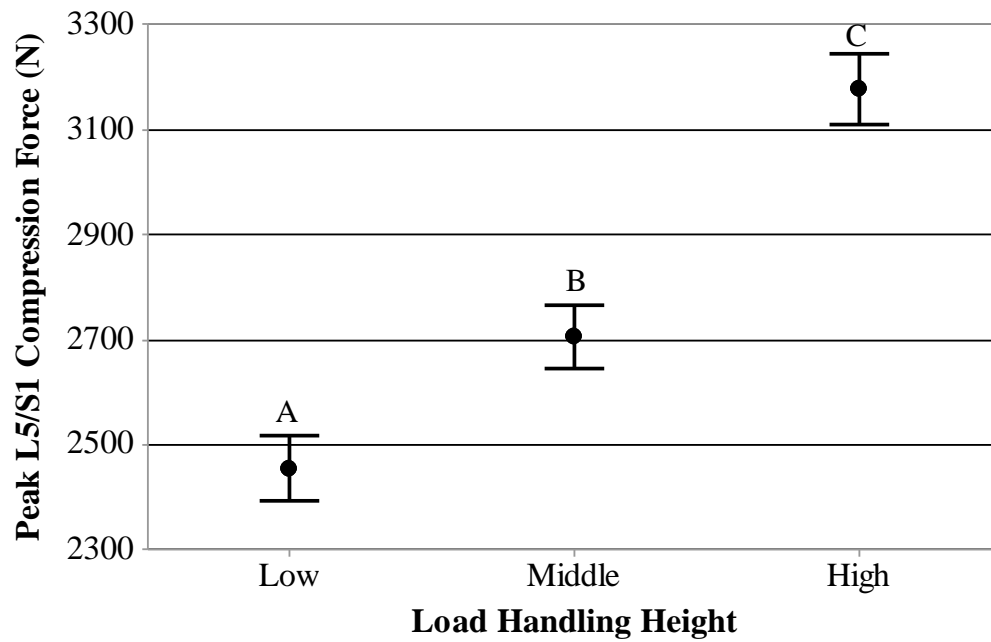


Figure 9. Peak L5/S1 compression force at three different load handling heights

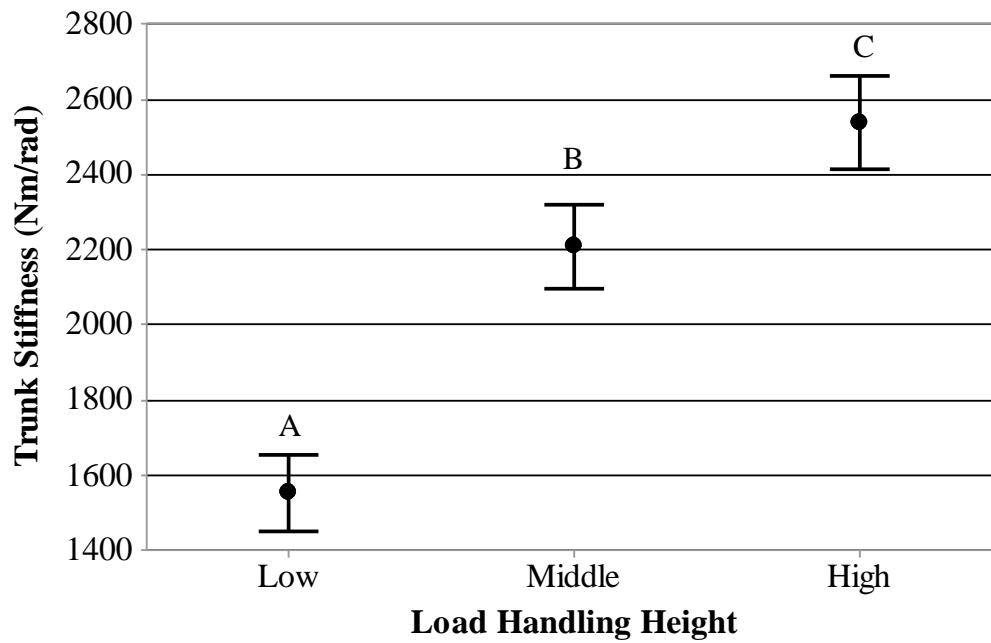


Figure 10. Trunk stiffness at three different load handling heights

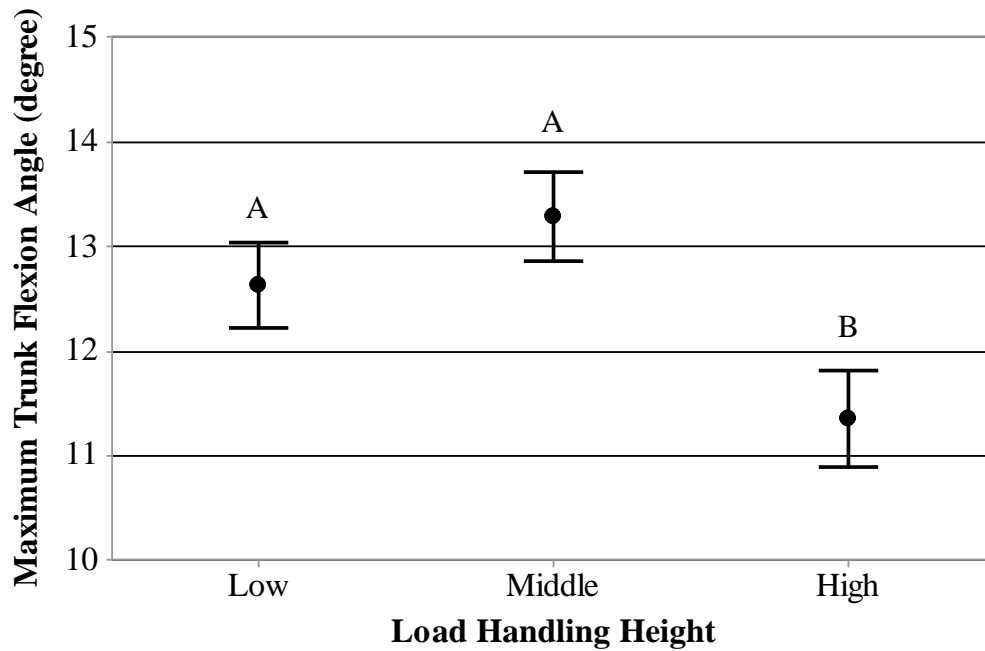


Figure 11. Maximum trunk flexion angle at three different load handling heights

Figures 12-14 respectively depict peak L5/S1 compression force, trunk stiffness, and maximum trunk flexion angle caused by sudden loading under the two different load configurations (three HEIGHT levels combined). Bars indicate the corresponding standard error.



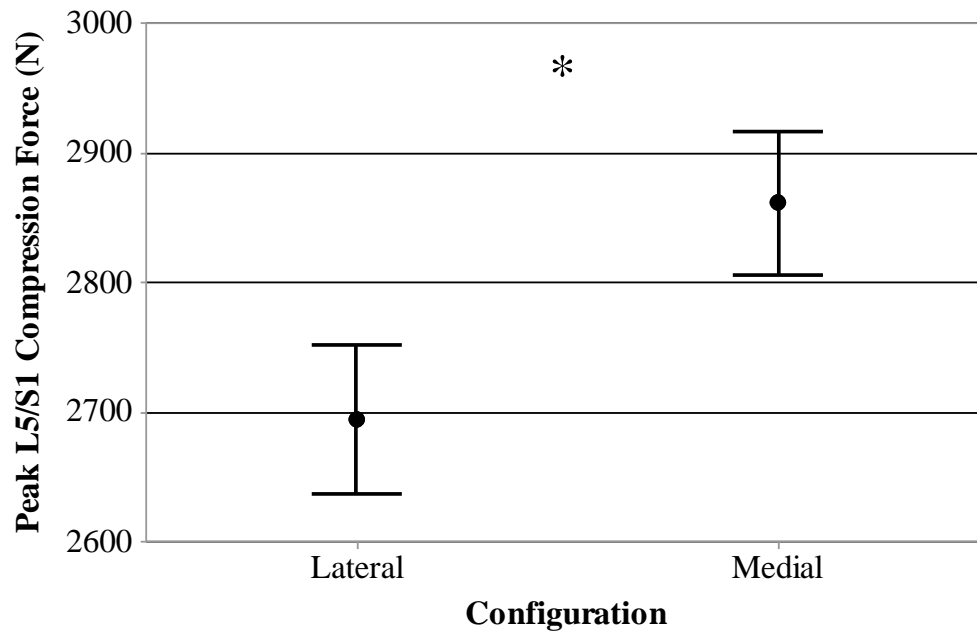


Figure 12. Peak L5/S1 compression force at two different load configurations. Asterisk denote statistical difference

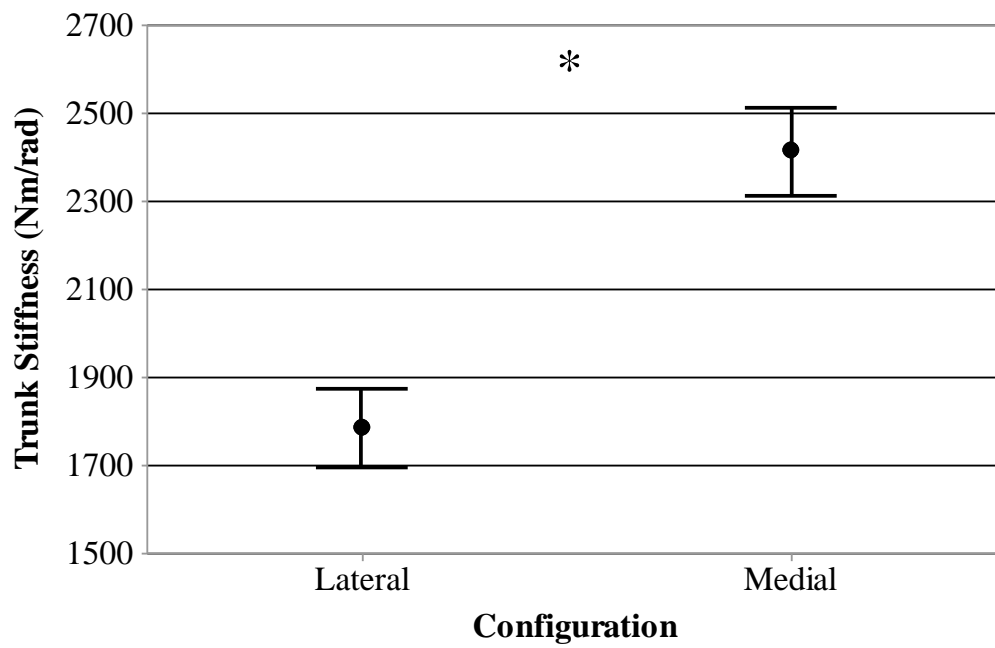


Figure 13. Trunk stiffness at two different load configurations. Asterisk denote statistical difference

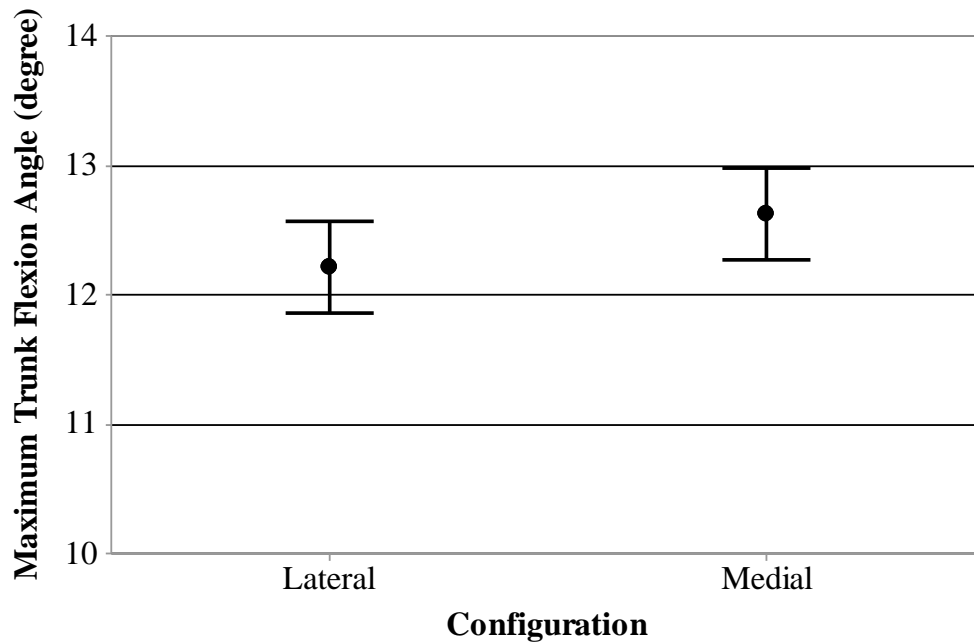


Figure 14. Maximum trunk flexion angle at two different load configurations. No significant effect was observed

#### 4.4. Discussion

The objective of this study was to investigate whether different weight configurations of the hand load could influence spinal loading and trunk stability during sudden loading incident. As described earlier, different weight configurations of a hand load create different rotational moments of inertia. Such differences could result in changes in neuromuscular control and muscle activation patterns in order to cope with external perturbations (i.e. sudden loading).

We hypothesized that during the performance of sudden loading when experiencing the same amount of external flexor moment, the weight with a smaller moment of inertia (i.e. medially distributed weight) may undergo larger angular accelerations and would compromise the stability of the spine and consequently elevate the mechanical loadings of the spinal column. Confirming

our initial hypotheses, findings of the present study demonstrated increased spinal compression force and trunk stiffness when handling medially distributed weight. As discussed earlier, when experiencing a medially distributed weight, the moment of inertia of the load decreases (Eq. 1) and as a result of this reduction, under the equal perturbations, it would undergo larger angular acceleration compared to the laterally distributed weight (Eq. 2). Thus, the trunk musculature is stiffened through muscle contraction in order to stabilize the load with larger displacements and bring it back to its equilibrium [38].

The present study estimated trunk stiffness from trunk kinematic response after the sudden load release. Since the involuntary muscle reflexes may have the potential to reinforce trunk kinematics, we referred to these estimates of stiffness as the effective trunk stiffness. According to the results of this experiment, a significantly higher effective trunk stiffness was observed when experiencing medially distributed weight (Fig. 13). Higher stiffness of the lumbar spine can be achieved by co-activation of agonistic and antagonistic trunk muscles which can lead to increased spinal stability [24, 25, 55, 94]. However, this enhancement in trunk stiffness and stability resulted in increased spinal loading which may elevate the risk of back injury [58]. Granata and Marras showed co-contraction of trunk muscles contributes to improved spinal stability and increased spinal compression, yet the stability improvement overweighs the increase of compression loads [38]. Moreover, results of the current experiment revealed no significant effect of weight configuration on maximum trunk flexion angle. However, there is a slight increase of maximum trunk flexion angle in medially distributed weight condition (on average, lateral:  $12.2^\circ$  and medial:  $12.6^\circ$ ). This trend is in agreement with previous literature which indicated improved spinal stability and stiffness with increased trunk flexion angle [24, 38].

In agreement with previous studies [49, 50], higher trunk stiffness and spinal compression force were observed when holding a load in a higher position. It has been well-documented that this increase in stiffness of torso musculature is to compensate for the reduced trunk and spinal stability [38, 49]. Also, the general decreasing trend of the maximum trunk flexion angle at a higher load handling position is in agreement with previous research [50]. Brown and colleagues demonstrated that under the same amount of external flexor moment, peak trunk flexion angle would drop as the stiffness of the lumbar region increases [95].

Finally, some limitations of this study need to be noted: first, stiffness and damping coefficients were assumed to be constant over the span of the data segment used to estimate these coefficients. Indeed, muscle reflex response alters its activation level and consequently its viscoelastic properties (i.e. stiffness and damping). Hence, these coefficients have been referred as effective trunk stiffness and damping. Another assumption that has been made was that the upper body (trunk, head, neck, and arms) was considered to be a rigid structure. It can be argued that although there was a deflection in elbow and shoulder joints throughout the aforementioned data segment, their stiffness and damping coefficients would appear in trunk stiffness coefficient. As a matter of fact, our results demonstrated higher trunk stiffness values compared to previous research, even though our weights were considerably lighter [71]. Another limitation of this study was that due to the relatively small trunk flexion during sudden loading, only spinal compression force was evaluated. More comprehensive spinal loading (shear force and torsional force) should be assessed in future studies especially when experiencing larger trunk posture deflections. Moreover, only one level of weight and one specific hand and trunk posture were tested in the current study, heavier weights and more complex hand and body postures (e.g. asymmetric weight holding) are recommended for future studies. Lastly, only young, male participants from the

student population were recruited in this experiment. The response of older individuals, female participants, or experienced workers warrants future investigation.

## CHAPTER 5. Conclusion

The results of the present study provided important information regarding the impact of static and sudden loading on trunk biomechanical responses when handling loads with different configurations (or distributions). These findings demonstrated that changes in the weight configuration of the hand load significantly influenced trunk muscle activities and spinal stability during the performance of manual material handling. According to our results, it was concluded that when confronted with static and sudden loading incidents, the load with a larger moment of inertia (i.e. laterally distributed load) could help reduce the risk of low back injury compared to the load with a smaller moment of inertia (i.e. medially distributed load).

Manual material handling tasks are prevalent in some industries such as construction, service, and transportation. During the performance of manual handling tasks (e.g. construction workers carry cement bags, baggage handlers remove bags and suitcases from conveyer belt, couriers deliver goods in boxes), sudden load release, heavy, repetitive and/or prolonged load handling due to loss of control and external impact can lead to back injuries [52, 54, 96, 97]. Results of this study suggested that to carry the same amount of weight, a symmetrical and more laterally distributed load helps improve spinal stability and reduce spinal loading as compared to a more medially distributed load.

Findings from the current study can be applied in numerous work scenarios. For instance, when construction workers carry cement bags, to enhance trunk stability and reduce spinal loading

these bags should be placed horizontally (instead of vertically) and in a close to abdomen position. Similarly, in air transportation, when a baggage handler carries baggage with both hands, holding handles along the long axis of the bag and keeping it close to the abdomen would help reduce the risk of spinal injury.

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## APPENDICES

### Appendix A: IRB Approval



Human Research Protocol  
Only Minimal Risk Consent Form  
Without HIPAA

#### Only Minimal Risk Consent Information Form (without HIPAA)

Principal Investigator	Ning, Xiaopeng
Department	ENGINEERING-Ind./Mgt. Sys. Engineering
Protocol Number	1508813734
Study Title	Biomechanical Assessment of Load Distribution on the Spine
Co-Investigator(s)	Madinei, Seyed Saman
Sponsor (if any)	None

#### Contact Persons

In the event you experience any side effects or injury related to this research, you should contact Dr. Xiaopeng Ning at (304) 294-9474. (After hours contact: Dr. Xiaopeng Ning at 515/520-1951). If you have any questions, concerns, or complaints about this research, you can contact Dr. Xiaopeng Ning at 304/294-9474. For information regarding your rights as a research subject, to discuss problems, concerns, or suggestions related to the research, to obtain information or offer input about the research, contact the Office of Research Integrity & Compliance at (304) 293-7073.

In addition if you would like to discuss problems, concerns, have suggestions related to research, or would like to offer input about the research, contact the Office of Research Integrity and Compliance at 304-293-7073.

#### Introduction

You, \_\_\_\_\_, have been asked to participate in this research study, which has been explained to you by Mr. Seyed Saman Madinei. This study is being conducted by Dr. Xiaopeng Ning (PhD), Seyed Saman Madinei in the Department of Industrial and Management System Engineering at West Virginia University.

#### Purpose(s) of the Study

The objective of the current study is to: 1) better understand the influence of holding weights with different load distributions on spine biomechanics; 2) better understand the influence of different load distributions on the stability and muscle activities of spine; 3) provide guidelines or recommendations for future biomechanical modeling and ergonomic software.

#### Description of Procedures

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Subject's

Initials \_\_\_\_\_  
Date \_\_\_\_\_

Approved:16-May-2016Expires:15-May-2017Number:1603051426

A) Upon arrival, experimental procedures and equipment will be first explained to you, and then a five minute warm-up session will be provided to allow you stretch your muscles and become familiar with the experimental setup. B) The non-invasive EMG electrodes will be placed using double-sided tape on your skin (EMG electrode, i.e. Electromyography electrode is a **non-invasive** skin sensor which receives the electrical responses of muscles while they contract). C) You will be instructed to complete the maximum voluntary contraction (MVC) trials for all muscles tested in the study. For each targeted muscle, you will perform two repetitions of MVC trials and each MVC trial lasts 6 seconds. One minute rest periods will be provided between MVC trials to ensure fatigue will not present for you. D) After completing the MVC trials the 3-D motion capture markers will be placed using double-sided tapes. E) After the 3-D motion capture markers placed on your skin, you will be directed where to stand and instructions on how to hold the weights and complete the experiment trials will be provided. F) You will be asked to conduct 2 different experiments. At the first one you will be requested to stay in a position of gripping a 20-lbs. lifting bar with 2 different load distributions when your eyes are closed. Then while the assistant guy is holding the weight with ropes, he will suddenly release the load and you will be required to grip it. When you gripped the load you have to lift the weight to its initial height. There will be 12 trials for this experiment each of which will take 5 seconds. G) Within the second experiment you will be asked to hold 20-lbs. weights with 2 different load distributions including concentrated and horizontally distributed load in front of your body for each trial. For the latter experiment there will be 12 trials each of which will take 10 seconds. The whole estimated time for the 2 experiments will be about 90 minutes (including the preparation time). H) Between every trial there will be one minute rest period in order to avoid fatigue in the muscles. Trials will be started upon command from the researcher conducting the experiment. The 3-D motion and EMG data will also be collected during the experimental trials.

### **Discomforts**

According to ergonomics assessments holding 20-lbs. weight for less than 10 seconds is considered as minimal risk of discomfort in low back.

### **Alternatives**

You do not have to participate in this study.

### **Benefits**

You may not receive any direct benefit from this study. The knowledge gained from this study may eventually benefit others.

### **Financial Considerations**

You will not receive any compensation for participation in the study

### **Voluntary Compensation**

You will not incur any costs related to the study. It is very important for you to understand that neither the investigator nor WVU or it associated affiliates has the funds set aside to pay for the cost work wages or any care or treatment that

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Subject's

Initials \_\_\_\_\_  
Date \_\_\_\_\_

**Approved:16-May-2016Expires:15-May-2017Number:1603051426**

might be necessary because you get hurt or sick taking part in this study. Any injuries that may result from this study would not be eligible for workers' Compensation as this is not a job related injury. Understand that any treatments necessary will be billed to the participant or to your personal health insurance, and you may wish to consult your insurance provider before participating in this study.

### **Confidentiality**

Any information about you that is obtained as a result of your participation in this research will be kept as confidential as legally possible.

In any publications that result from this research, neither your name nor any information from which you might be identified will be published without your consent.

### **Voluntary Participation**

Participation in this study is voluntary. You are free to withdraw your consent to participate in this study at any time.

Refusal to participate or withdrawal will not affect [your class standing or grades, as appropriate] and will involve no penalty to you. Refusal to participate or withdrawal will not affect your future care, or your employee status at West Virginia University.

In the event new information becomes available that may affect your willingness to participate in this study, this information will be given to you so that you can make an informed decision about whether or not to continue your participation.

You have been given the opportunity to ask questions about the research, and you have received answers concerning areas you did not understand.

Upon signing this form, you will receive a copy.

I willingly consent to participate in this research.

### **Signatures**

Signature of Subject

Printed Name

Date

Time

The participant has had the opportunity to have questions addressed. The participant willingly agrees to be in the study.

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Subject's

Initials \_\_\_\_\_  
Date \_\_\_\_\_

**Approved:16-May-2016Expires:15-May-2017Number:1603051426**

Signature of Investigator or Co-Investigator

Printed Name

Date

Time



## Appendix B: Equations for Trunk Muscles' Moment Arms

$ESMomArmRx = 6 - 5.7 * (Twidth / Mb)$ ; % (cm) L5 level coronal plane moment arm for right ES

$ESMomArmLx = 5.7 - 4.42 * (Twidth / Mb)$ ; % (cm) L5 level coronal plane moment arm for left ES

$ESMomArmRy = 2.73 + 0.023 * Hb * Mb$ ; % (cm) L5 level sagittal plane moment arm for right ES

$ESMomArmLy = -1.83 + 4.2 * Hb$ ; % (cm) L5 level sagittal plane moment arm for left ES

$MUMomArmRx = 1.1$ ; % (cm) coronal plane moment arm for right MU

$MUMomArmLx = 1.4$ ; % (cm) coronal plane moment arm for left MU

$MUMomArmRy = 5.5$ ; % (cm) sagittal plane moment arm for right MU

$MUMomArmLy = 5.3$ ; % (cm) sagittal plane moment arm for left MU

$EOMomArmRx = 1.15 + 6.6 * Hb$ ; % (cm) coronal plane moment arm for right EO

$EOMomArmLx = 4.84 + 0.156 * Mb / Hb$ ; (cm) coronal plane moment arm for left EO

$EOMomArmRy = -13.8 + 1.267 * Tdepth / Hb$ ; % (cm) sagittal plane moment arm for right EO

$EOMomArmLy = -13.8 + 1.267 * Tdepth / Hb$ ; % (cm) sagittal plane moment arm for left EO

$RAMomArmRx = -5.44 + 0.278 * 1.2 * Twidth$ ; % (cm) L5 level coronal plane moment arm for right RA

$RAMomArmLx = -5.44 + 0.278 * 1.2 * Twidth$ ; % (cm) L5 level coronal plane moment arm for left RA

$RAMomArmRy = -5.52 + 1.033 * (Tdepth / Hb)$ ; % (cm) L5 level sagittal plane moment arm for right RA

$RAMomArmLy = -3.03 + 0.46 * Tdepth$ ; % (cm) L5 level sagittal plane moment arm for left RA

## Appendix C: Equations for Trunk Muscles' Cross Sectional Area

ESrCSA=21.15; %CSA for right ES

ESlCSA=21.77; %CSA for left ES

MUrCSA=4.47; %CSA for right MU

MUICSA=4.72; %CSA for left MU

RArCSA=-1.392+0.131\*Mb; %CSA for right RA

RAlCSA=-1.93+0.137\*Mb; %CSA for left RA

EOrCSA=1.2+0.013\*Tdepth\*Twidth; %CSA for right EO

EOlCSA=1.2+0.013\*Tdepth\*Twidth; %CSA for left EO